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Pediatric CT: Strategies to Lower Radiation Dose

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Abstract

OBJECTIVE—The introduction of MDCT has increased the utilization of CT in pediatric radiology along with concerns for radiation sequelae. This article reviews general principles of lowering radiation dose, the basic physics that impact radiation dose, and specific CT integrated dose-reduction tools focused on the pediatric population.

CONCLUSION—The goal of this article is to provide a comprehensive review of the recent literature regarding CT dose reduction methods, their limitations, and an outlook on future developments with a focus on the pediatric population. The discussion will initially focus on general considerations that lead to radiation dose reduction, followed by specific technical features that influence the radiation dose.

Keywords

CT; pediatric; radiation

The number of CT examinations in the United States has significantly increased in recent decades. Approximately 62 million CT scans are performed each year [1], with 4–7 million of those performed on children [2]. According to Mettler et al. [1], CT examinations account for only 15% of the total number of procedures but for over half of the collective dose, when dental scans are excluded. These numbers are based on a report from the National Council on Radiation Protection and Measurement [3]. According to Brenner and Hall [4], the largest increase in CT use has been in the pediatric population and in adult screening.

The advent of MDCT at the 1998 meeting of the Radiological Society of North America was a significant step forward toward isotropic volume imaging [5]. This changed the way radiologists look at CT images. From then on, the routine use of multiplanar reformations in the sagittal and coronal plane and detailed cardiac imaging became feasible for the first time. New indications for CT emerged. The faster gantry rotation cycle times markedly reduced the need for anesthesia in children [4], which led to increased utilization of this powerful diagnostic tool in the pediatric population.

In 2001, an article in *USA Today* [6] based on two articles published in *American Journal of Roentgenology* [7, 8] and one article published in the *Journal of Radiologic Protection* [9] raised the public's awareness that routine CT scans have the potential to cause fatal cancers in children and that, in general, children receive a much higher radiation dose than is necessary. Prompted by public concerns, the Society for Pediatric Radiology organized a multidisciplinary conference to address the concerns of CT radiation exposure in children [10]. The ALARA (as low as reasonably achievable) concept, which was developed in the 1960s by the Atomic Energy Commission (the predecessor of the Nuclear Regulatory Commission) and the Department of Energy, was strongly embraced as a unifying concept for pediatric CT dose reduction efforts [11].

At the end of 2002, there were still relatively few guidelines established for performing MDCT in children. Until then, very little attention was paid to adjusting the scanning parameters, such as beam energy (kVp), tube current (mA), and pitch, to the patient's size (i.e., sized-based scanning) [12]. At the end of 2003, Frush et al. [13] wrote, "Despite the increase in use and increased attention, there has not been a parallel increase in understanding of the risk or the use of techniques for reducing these risks."

In 2008, the *Image Gently* campaign [14] was founded by the Society for Pediatric Radiology, the American Society of Radiologic Technologists, the American College of Radiology, and the American Association of Physicists in Medicine (AAPM) with the goal of increasing pediatric CT radiation dose awareness by starting a national education and awareness program (Fig. 1).

Although the risk of experiencing radiation-induced cancer for a given child is small, the concern is related to the rapid increase in the use of radiation for diagnostic purposes. The rapidly dividing cells in children are more radiosensitive than those of adults [2]. In addition, children have a longer lifetime in which radiation-related cancers can become manifest, and the current conservative theory is that the cancer risk is cumulative over a lifetime [13]. This means that each x-ray examination contributes to the lifetime exposure and therefore increases the risk of inducing a fatal cancer.

A publication of the National Cancer Institute in 2008 showed that the effective dose for an unadjusted abdominal CT (200 mAs) of a child is 11–24 mSv; the effective dose is 3–6 mSv when an adjusted (50 mAs) scan protocol is used [15]. The adjustment of scan parameters was driven by the patient's weight [15]. Approximately one third of children who have undergone CT scans have already had at least three different scans in their lifetime added to their cumulative dose [15].

There is no doubt that CT is an extremely important imaging tool for the pediatric population [13] and that it has saved many lives. Clearly, there are limits to lowering radiation dose. Lowering the risk of radiation-induced cancer against the risk of missing the correct diagnosis because of poor image quality has to be put into perspective [16].

Recent studies show that, during the last several years, the number of CT scans performed on children has decreased. This is in particular true for children's hospitals [17–19].

At present, there are no regulations in place to track cumulative patient radiation dose [20]. Work is under way to change this on an international level through the International Atomic Energy Agency (Smartcard Project) [21], and numerous vendors are beginning to offer dose-tracking software to help monitor radiation dose.

General Considerations Regarding CT Dose Reduction

According to Strauss et al. [2], technologists trained before 2007 were not trained in the physics of CT equipment. Increasing the awareness and the understanding of CT radiation dose issues among doctors and technologists will lead to improvement of scan protocols and therefore reduce radiation dose [2]. The *Image Gently* website [22] offers free online technologist education modules.

For this reason, postprimary CT certification through the American Registry of Radiologic Technologists and accreditation from the American College of Radiology should be encouraged [14]. During the accreditation process, radiologists, medical physicists, and radiologic technologists learn about their equipment and determine whether their protocols are conforming to the national guidelines [2].

The *Image Gently* campaign also recommends enlisting the service of a qualified physicist, who ensures that the technical aspects of CT are well understood and are applied to the unique design of the facility's CT scanners [2]. The medical physicist should be certified by the American Board of Radiology or the American Board of Medical Physics in a diagnostic imaging physics subject area or in radiologic physics [2].

It is also imperative to use alternative imaging methods that do not use ionizing radiation whenever feasible [2]. For example, instead of using CT, rapid single-sequence MRI could be used to rule out ventriculoperitoneal shunt malfunction as described by Ashley et al. [23] in 2005 or cranial ultrasound in younger patients with open fontanelles [24]. Increased sedation time and other risks from MRI acquisitions, however, may outweigh the risks of CT in some situations [25].

To avoid unnecessary CT examinations, it is first crucial to evaluate whether the ordered CT is justified in every single case [2]. If, after preliminary review, CT proves to be the appropriate method then, according to Strauss et al. [2], single-phase CT scans are usually all that is needed in children and one long scan results in a lower radiation dose than several regional scans overlapping each other at the scan end and start.

It is useful to provide imaging consultations and training on radiation protection for referring physicians to help them to find the appropriate diagnostic pathway [2]. This speeds up the decision-making process and eases communication. There is a free Microsoft Power-Point presentation on the *Image Gently* website [22] on radiation protection for children for this purpose. This will help to reduce the number of nonindicated CT requests.

In particular, for chronic diseases, imaging modalities that do not require the use of ionizing radiation should be used whenever feasible to keep the cumulative lifetime radiation dose as

low as possible. For example, MRI enterography should be considered in inflammatory bowel disease as an alternative diagnostic method [2].

CT Dose Metrics

The CT dose index (CTDI) is a measure of the absorbed dose to a standard plastic phantom. The weighted CTDI ($CTDI_w$) is calculated as the weighted average of the absorbed dose to the center and periphery of a phantom (in an attempt to account for nonuniform dose distributions in a patient slice). The volume CTDI ($CTDI_{vol}$) is the current industry standard for reporting absorbed dose from a CT acquisition and is essentially the $CTDI_w$ normalized by pitch, to account for potential nonuniform axial dose distributions from multislice helical acquisitions [26]. It is measured in two standard cylindrical phantoms of 16 cm diameter (head phantom) and 32 cm diameter (body phantom) and is reported in units of grays. Finally, the $CTDI_{vol}$ represents the average absorbed dose to a single slice. To account for dose to a scan range, this measure is multiplied by the scan range (centimeters), to calculate the dose-length product (DLP in milligrays times centimeters).

The $CTDI_{vol}$ and DLP are measures of dose to a plastic phantom and are indexes that allow comparison between scan protocols, but they should not be confused with patient dose or biologic risk. It is often clinically beneficial to represent dose from CT with a measure that is useful for comparison with other ionizing radiation modalities. A common quantity for comparing doses is effective dose. This construct was proposed by the International Commission on Radiation Protection [27], and, despite its significant limitations [28], effective dose is often used as a convenient measure to describe radiation doses from imaging procedures. In brief, effective dose represents the potential risk of cancer from nonuniform radiation exposures based on individual organ doses and organ sensitivities [13]. Effective dose can be estimated from CT acquisitions through simulation data that relate DLP values to organ-specific doses. Then, the weighted average of organ-specific doses times their relative risk of forming cancers provides estimates of effective dose.

Effective dose estimation from $CTDI_{vol}$ currently is based on fairly simplistic models, which can underestimate the dose levels in pediatric patients. In 2011, the AAPM task group 204 [29] published a report proposing a new measure—size-specific dose estimates—for conversion of $CTDI_{vol}$ and DLP to patient dose. This new measure uses conversion factors based on size-appropriate patient models, leading to more accurate dose measures in the range of patient sizes in the pediatric population.

Image Quality Versus Dose

Determining the necessary image quality and dose to provide sufficient task-based performance is often very challenging [18]. There are phantom acquisitions that can provide objective assessment of image performance of conventional image metrics, such as spatial resolution, contrast resolution, image uniformity, and CT number accuracy. In general, these metrics are useful for characterizing system performance but are rarely used for selection of patient protocols. In clinical CT, images are required to perform a range of tasks, from detection of small features to lesion characterization. Because of this range of tasks, it is challenging, if not impossible, to determine a fixed set of acquisition parameters that will

provide “necessary” image quality. Moreover, because of the wide range of clinician preferences and experience, necessary image quality varies. For these reasons, local protocols are developed on the basis of local experience.

As with all processes at an institution, CT protocols should be reviewed regularly to ensure that image quality and dose are being optimized. If all clinicians view the current protocols as having sufficient image quality, these can be considered as baseline levels. Phantom acquisitions using baseline levels could be compared with successive reductions in image quality or dose. Often, minor dose reductions on the order of 5–15% lead to acceptable increases in noise [2]. Determining these noise increases in phantom studies may provide evidence for implementing minor dose reductions in a subset of patient studies (e.g., testing minor reductions in patient follow-up studies may be more appropriate than attempting reductions in initial staging studies). If minor reductions are found to be acceptable in all cases of the subset of patients, these new protocols may be appropriate for all patients. This graduated systematic evaluation of dose reductions and image quality tolerance can lead to more dose-conscious CT. It should be stressed that over- and underdosing are medical errors and, therefore, dose reductions and dose increases may be appropriate while trying to optimize CT protocols.

Acquisition Parameters to Reduce CT Radiation Dose

To be able to use the scanning protocols effectively, one has to understand the impact of the single-scan parameters with respect to radiation dose. Parameters that can be manipulated and that have a direct influence on the radiation dose include x-ray beam energy (measured in kilovoltage peak), tube current (measured in milliamperes), gantry rotation time (equals the exposure time), section thickness (also called collimation), pitch (defined as table distance traveled in one 360° gantry rotation), distance from the x-ray tube to the CT isocenter [26], and the scan length [30] (Table 1).

The relationship between beam energy and radiation dose is nonlinear; for example, changing the beam energy from 80 to 100 kVp will change the CTDI in a head phantom from 14 to 26 mGy (Fig. 2). The relationship between the tube current and the radiation dose is linear, which means that increasing the tube current by 50% will result in a 50% higher dose [26] (Fig. 3). Tube current (in milliamperes) and gantry rotation time (in seconds) are often coupled. In these cases, the milliamperere-second value has a linear relationship with the resulting radiation dose [26].

Pitch and radiation dose are inversely proportional. Scans with a pitch of 2 give 50% of the radiation doses of scans with a pitch of 1 [32] (Fig. 4). However, some scanners automatically link pitch and tube current such that, as the pitch increases, the tube current increases proportionally, preventing radiation dose reduction. In this setting, an increased pitch just leads to increased scan speed.

Single-detector and MDCT scanners both show an increase in CTDI values if thinner total beam widths are chosen (Fig. 5). This is primarily because the x-ray beam is always slightly wider than the axial length of the detectors (beam has an axial penumbra). When performing an acquisition through some axial length, smaller collimated beams will require more

overlap of the penumbra, meaning that more of the x-ray flux is not contributing to the image; it is merely increasing the dose. This effect is called “overbeaming” and is shown in Figure 6A. The impact of choosing a thin total beam width using an MDCT scanner is much more significant in terms of increase of radiation dose, compared with a single-slice scanner [26]. The collimation of single-slice scanners is particularly optimized for the single-slice geometry. Another source of excess radiation arises from helical acquisitions, in which roughly a one-half rotation of the gantry is required at the beginning and end of the scan range to provide sufficient data for reconstruction of the first and last slice, as illustrated in Figure 6B. This effect, known as overranging or overscanning, results in some wasted dose for collecting slices that are not contributing to the reconstructed first slice [32]. In general, the dose from overranging increases along with increases in pitch and detector collimation (Fig. 6).

Positioning the patient in the middle of the CT gantry reduces the radiation dose to the patient according to the inverse-square law [2, 18]. An MDCT phantom study conducted by Li et al. [33] found that peripheral and surface CTDI values increased by approximately 12–18% and 41–49% when the phantom was 30 and 60 mm off center, respectively. Off-centering also leads to decreased image quality with increased image noise.

The extent of the scout and the scan should be limited to the area of concern. Strauss et al. [2] recommended reducing the radiation dose for the topogram and changing the orientation of the topogram from anteroposterior to posteroanterior in a supine patient if the computer tomograph allows it. This reduces the dose to male gonads, breast, thyroid, and eye lenses [2]. By reducing the x-ray beam energy from 120 to 80 kVp and changing the tube position to posteroanterior orientation, the radiation dose could be less than that of a chest-x-ray [2].

To lower radiation dose with the bolus-tracking method, Goo [34] suggested monitoring just slightly before the expected occurrence of the contrast peak in the targeted vessel.

Scanner Integrated Radiation Saving Tools

Automatic Exposure Control

Automatic exposure control (AEC) refers to different methods of adapting the CT tube current to the patient attenuation of the x-ray beam. Depending on the CT manufacturer, different methods and vendor-specific names are used. For example, GE Healthcare Technologies has a system called Smart mA, Philips Healthcare uses Z-DOM and D-DOM, Siemens Healthcare calls its system CareDose 4D, and finally Toshiba Medical Systems calls its system Sure Exposure [35]. According to McCollough [36], modulation angularly around the patient and along the z-axis is optimal. The precondition is that the tube current is adapted to the patient size [36] (Fig. 7).

An investigation of Greess et al. [37] showed that the use of MDCT online tube current modulation in children led to a radiation dose reduction from 26% to 43%, depending on the child's geometry and weight, without compromising image quality.

Peng et al. [38] used an automatic tube current modulation method with a standardized noise index for chest CT in young children (0.2–3 years). The scanner automatically selected the

actual tube current according to the noise index value, which was determined by the examiner. The image quality was sufficient in all studies. The radiation dose index (CTDI_{vol}) of the study group compared with the control group with a fixed tube current was about 65% lower [38].

Adaptive Section Collimation

Adaptive section collimation, which is available in some new scanners, is a method to reduce radiation dose due to overscanning or overranging in the z-axis and is particularly effective in scan ranges smaller than 12 cm. The dose savings are up to 38% [39]. This dose-reduction method especially benefits very young children.

Bowtie Filters

Bowtie filters are CT filters that harden the x-ray beam by removing all of the low-energy x-rays that would otherwise be absorbed by the patient and not reach the detector. They also concentrate the x-rays in the central part of the scanned object. This leads to increased image quality and a 50% reduction in surface dose when compared with flat filters. The functionality of bowtie filters depends crucially on the proper positioning of the patient in the gantry isocenter [18].

Postprocessing Methods to Enable Reduced Dose

Factors that have an indirect influence on radiation dose are those that have an influence on the image generation, such as the reconstruction filter. For example, the CT operator might be able to achieve the same level of background noise in a lower dose 100-kVp image through use of a smoother reconstruction filter than a 120-kVp image with a standard reconstruction filter [26].

After acquisition, the raw CT data are reconstructed into images. The conventional method for performing this reconstruction is a variant of the filtered back projection algorithm. This analytic method requires a filtering step before reconstruction. All vendors offer several options for filtering the data. Essentially, these filters can reduce noise, usually at the expense of spatial resolution. This noise reduction could effectively enable a reduction in acquisition technique and dose at a matched noise level (while causing degradation in resolution). Some imaging tasks, in which resolution is not paramount, may warrant this simple strategy for noise reduction.

Numerous vendors are offering image enhancement methods to further reduce noise in images, enabling further dose reduction. These methods are often classified as iterative algorithms. They perform a prereconstruction iterative filtering process, a postreconstruction iterative filter, or a more thorough fully iterative reconstruction of the image. Initial studies of the pre- and postreconstruction filtering methods suggest radiation dose savings on the order of 10–40% at matched image quality. The fully iterative reconstruction methods have the potential to offer even more promising results [40–43].

Scanner-Independent Radiation Dose Saving Methods: Bismuth Shielding

Combining AEC and bismuth shielding for the thyroid gland and breast leads to further radiation dose savings. In an anthropomorphic phantom study representing a 5-year-old child using an age-based 16-MDCT scan protocol, Coursey et al. [44] showed a 52% radiation dose reduction using bismuth breast shielding when the shield was placed after the scout was performed. If the shield is placed before the scout is taken, the AEC compensates for the reduced x-ray penetration, thereby increasing the dose to the patient, counter to the desired outcome. In another anthropomorphic study using a standardized cervical spine trauma protocol for adults and a thyroid as well as breast bismuth shielding, Gunn et al. [45] found a 22.5% dose reduction to the thyroid and a 36.6% radiation dose reduction to the breast if the shielding was placed directly onto the skin. If the shielding was placed on an immobilization collar, the radiation dose reduction was not statistically significant. They used a 64-MDCT scanner with AEC. However, in February 2012, the AAPM released a statement regarding bismuth shielding in combination with AEC emphasizing that incorrect use leads to increased radiation dose [46]. On some scanners, even the placement of the bismuth shielding after the scout was taken (correct method) can lead to increased radiation dose because the scanner is able to adjust to the increased attenuation during the scan. Other side effects are streak and beam-hardening artifacts. The Society of Cardiac CT discourages the use of bismuth shielding during coronary calcification measurements to avoid inaccurate results [47]. Instead of using bismuth shielding, the AAPM recommends working with the medical physicist or CT application specialist to optimize the AEC parameters [46].

Cardiac CT in the Pediatric Population

One of the main challenges of performing pediatric ECG gated cardiac CT is the high heart rates encountered in very young children, which requires cardiac-gated acquisitions. Relentless improvements in CT hardware with increasing temporal and spatial resolution and radiation dose–reduction methods have made it possible to use CT in children with congenital heart disease.

There are three principal methods currently used for imaging the heart and great vessels, broadly categorized as non-ECG-synchronized helical CT scan, retrospectively ECG-gated spiral scanning, and prospectively ECG-triggered sequential scan. The non-ECG-synchronized scan is often used to examine the heart and the origin of the great vessels in patients with congenital heart disease. According to Goo [34], in 82% of patients, even the origin and the proximal segment of the coronary arteries are visualized and the effective radiation dose is usually less than 1.0 mSv if the scan parameters are adequately adjusted to the patient weight and if modern dose-saving tools, such as adaptive section collimation, are applied.

Retrospectively ECG-gated helical CT allows both morphologic and functional evaluation of the heart. In general, the radiation dose is much higher compared with a non-ECG-synchronized helical CT scan and a prospectively ECG-triggered sequential scan because of the low pitch [34].

One important method to reduce radiation dose in retrospectively ECG-gated helical CT is the use of ECG-controlled tube current modulation. The examiner can choose a predefined time window during the cardiac cycle when the tube current is reduced to either 20% or 4% of the original tube current. Functional evaluations are excluded if the tube current is reduced to 4% [34]. Ideally, this can save up to 64% of the radiation dose [34]. According to Goo [34], the effective dose estimates of a retrospectively ECG-gated dual-source CT should not be more than 2–6 mSv in congenital heart disease (Fig. 8).

Prospectively ECG-triggered sequential CT (step-and-shoot method) has the lowest radiation dose. According to Goo [34], it ranges from 0.2 to 0.7 mSv in newborns and infants. Even if a non-breath-hold technique is used, the images are often motion free [34]. Disadvantages of this method include the lack of functional assessment because of the limited data acquisition during only a portion of the full cardiac cycle [34] and stairstep artifacts related to the step-and-shoot mechanics [48].

Paul et al. [49] reported even lower effective radiation dose measurements. Their values ranged between 0.05 and 0.8 mSv. They evaluated the step-and-shoot method in infants and small children with congenital heart disease using a dual-source 128-MDCT scanner. The scan protocol consisted of an 80-kVp tube voltage and a tube current that was adjusted to the body weight as follows: 10 mAs/kg up to 6 kg and then 5 mAs/kg up to 90 mAs.

Conclusion

The rapid technical development of radiation-sparing techniques in the last 10 years clearly shows the increased radiation dose awareness among CT scanner manufacturers. Continuous research and innovation will lead to even more and better dose-saving tools. Therefore, it is important to stay informed and to keep current with the newest technology to offer the best available service with the lowest radiation burden to our patients. According to Goo [34], the future will bring a high-pitch (up to 3.4) dual-source CT and new types of tube current modulation with respect to the superficial tissue. Furthermore, automated scan protocols based on the clinical question and the anatomic region are being developed. Likewise, new image-enhancement methods improving the image reconstruction process offer the potential for significant dose savings at matched image quality.

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References

1. Mettler FA, Thomadsen BR, Bhargavan M, et al. Medical radiation exposure in the U.S. in 2006: preliminary results. *Health Phys.* 2008; 95:502–507. [PubMed: 18849682]
2. Strauss KJ, Goske MJ, Kaste SC, et al. Image Gently: ten steps you can take to optimize image quality and lower CT dose for pediatric patients. *AJR.* 2010; 194:868–873. [PubMed: 20308484]
3. National Council on Radiation Protection and Measurements. NCRP report no. 160: ionizing radiation exposure of the population of the United States. Bethesda, MD: National Council on Radiation Protection & Measurements; 2009.

4. Brenner DJ, Hall EJ. Computed tomography: an increasing source of radiation exposure. *N Engl J Med*. 2007; 357:2277–2284. [PubMed: 18046031]
5. Flohr, TH.; Klingensbeck-Regn, K.; Ohnesorge, B.; Schaller, S. Multislice scanning with the SOMATOM volume zoom. In: Marin ek, B.; Ros, PR.; Reiser, M.; Baker, ME., editors. *Multislice CT: a practical guide—proceedings of the 5th International SOMATOM CT Scientific User Conference*, Zurich, June 2000. Heidelberg, Germany: Springer-Verlag; 2001. p. 7-21.
6. Sternberg S. CT scans in children linked to cancer later. *USA Today*. 2012 Jun 19.
7. Brenner DJ, Elliston CD, Hall EJ, Berdon WE. Estimated risk of radiation-induced fatal cancer from pediatric CT. *AJR*. 2001; 176:289–296. [PubMed: 11159059]
8. Donnelly LF, Emery KH, Brody AS, et al. Minimizing radiation dose for pediatric body applications of single-detector helical CT: strategies at a large children's hospital. *AJR*. 2001; 176:303–306. [PubMed: 11159061]
9. Mettler FA Jr, Wiest PW, Locken JA, Kelsey CA. CT scanning: patterns of use and dose. *J Radiol Prot*. 2000; 20:353–359. [PubMed: 11140709]
10. The ALARA (as low as reasonably achievable) concept in pediatric CT intelligent dose reduction: multidisciplinary conference organized by the Society of Pediatric Radiology, August 18–19, 2001. *Pediatr Radiol*. 2002; 32:217–313. [No authors listed].
11. United States Nuclear Regulatory Commission. ALARA. [Accessed September 19, 2011] Nuclear Regulatory Commission website. www.nrc.gov/reading-rm/basic-ref/glossary/alara.html
12. Frush DP, Slack CC, Hollingsworth CL, et al. Computer-simulated radiation dose reduction for abdominal multidetector CT of pediatric patients. *AJR*. 2002; 179:1107–1113. [PubMed: 12388482]
13. Frush DP, Donnelly LF, Rosen NS. Computed tomography and radiation risk: what pediatric health care provider should know. *Pediatrics*. 2003; 112:951–957. [PubMed: 14523191]
14. Goske MJ, Applegate KE, Boylan J, et al. The 'Image Gently' campaign: increasing CT radiation dose awareness through a national education and awareness program. *Pediatr Radiol*. 2008; 38:265–269. [PubMed: 18202842]
15. National Cancer Institute at the National Institutes of Health. [Accessed September 19, 2011] Radiation risks and pediatric computed tomography (CT): a guide for health care providers. National Cancer Institute website. www.cancer.gov/cancertopics/causes/radiation/radiation-risks-pediatric-CT
16. Cohen MD. CT radiation dose reduction: can we do harm by doing good? *Pediatr Radiol*. 2012; 42:397–398. [PubMed: 22302318]
17. Slovis TL. Where we were, what has changed, what needs doing: a decade of progress. *Pediatr Radiol*. 2011; 41:456–460. [PubMed: 21847722]
18. Callahan MJ. CT dose reduction in practice. *Pediatr Radiol*. 2011; 41:488–492. [PubMed: 21847727]
19. Bachur RG, Hennelly K, Callahan MJ, Monuteaux MC. Advanced radiologic imaging for pediatric appendicitis, 2005–2009: trends and outcomes. *J Pediatr*. 2012; 160:1034–1038. [PubMed: 22192815]
20. Colang JE, Killion JB, Vano E. Patient dose from CT: a literature review. *Radiol Technol*. 2007; 79:17–26. [PubMed: 17848530]
21. Frush DP. CT dose and risk estimates in children. *Pediatr Radiol*. 2011; 41:483–487. [PubMed: 21847726]
22. Alliance for Radiation Safety in Pediatric Imaging. Image gently. Alliance for Radiation Safety in Pediatric Imaging website. www.imagegently.org Updated 2011.
23. Ashley WW Jr, McKinstry RC, Leonard JR, Smyth MD, Lee BC, Park TS. Use of rapid-sequence magnetic resonance imaging for evaluation of hydrocephalus in children. *J Neurosurg*. 2005; 103(suppl 2):124–130. [PubMed: 16370277]
24. Smyth MD, Narayan P, Tubbs RS, et al. Cumulative diagnostic radiation exposure in children with ventriculoperitoneal shunts: a review. *Childs Nerv Syst*. 2008; 24:493–497. [PubMed: 18180935]
25. Practice advisory on anesthetic care for magnetic resonance imaging: a report by the American Society of Anesthesiologists Task Force on Anesthetic Care for Magnetic Resonance Imaging. *Anesthesiology*. 2009; 110:459–479. [No authors listed]. [PubMed: 19237869]

26. McNitt-Gray MF. AAPM/RSNA physics tutorial for residents: topics in CT—radiation dose in CT. *RadioGraphics*. 2002; 22:1541–1553. [PubMed: 12432127]
27. The 2007 Recommendations of the International Commission on Radiological Protection: ICRP publication 103. *Ann ICRP*. 2007; 37:1–332. [No authors listed].
28. Borrás C, Huda W, Orton CG. The use of effective dose for medical procedures is inappropriate. *Med Phys*. 2010; 37:3497–3500. [PubMed: 20831055]
29. American Association of Physicists in Medicine. AAPM report no. 204: size-specific dose estimates (SSDE) in pediatric and adult body CT examinations. American Association of Physicists in Medicine website. www.aapm.org/pubs/reports/rpt_204.pdf. Published 2011.
30. Frush DP. Strategies of dose reduction. *Pediatr Radiol*. 2002; 32:293–297. [PubMed: 11956713]
31. Linton OW, Mettler FA Jr. National conference on dose reduction in CT, with an emphasis on pediatric patients. *AJR*. 2003; 181:321–329. [PubMed: 12876005]
32. Schilham A, van der Molen AJ, Prokop M, de Jong HW. Overranging at multisection CT: an underestimated source of excess radiation exposure. *RadioGraphics*. 2010; 30:1057–1067. [PubMed: 20631368]
33. Li J, Udayasankar UK, Toth TL, Seamans J, Small WC, Kalra MK. Automatic patient centering for MDCT: effect on radiation dose. *AJR*. 2007; 188:547–552. [PubMed: 17242267]
34. Goo HW. State of the art CT imaging techniques for congenital heart disease. *Korean J Radiol*. 2010; 11:4–18. [PubMed: 20046490]
35. McCollough CH, Bruesewitz MR, Kofler JMCT Jr. Dose reduction management tools: overview of available options. *RadioGraphics*. 2006; 26:503–512. [PubMed: 16549613]
36. McCollough CH. Automatic exposure control in CT: are we done yet? *Radiology*. 2005; 237:755–756. [PubMed: 16304094]
37. Greess H, Lutze J, Noemayr A, et al. Dose reduction in subsecond multislice spiral CT examination of children by online tube current modulation. *Eur Radiol*. 2004; 14:995–999. [PubMed: 15052502]
38. Peng Y, Li J, Ma D, et al. Use of automatic tube current modulation with a standard noise index in young children undergoing chest computed tomography scans with 64-slice multidetector computed tomography. *Acta Radiol*. 2009; 50:1175–1181. [PubMed: 19922316]
39. Deak PD, Langner O, Lell M, Kalender WA. Effects of adaptive section collimation on patient radiation dose in multisection spiral CT. *Radiology*. 2009; 252:140–147. [PubMed: 19561253]
40. Hu XH, Ding XF, Wu RZ, Zhang MM. Radiation dose of non-enhanced chest CT can be reduced 40% by using iterative reconstruction in image space. *Clin Radiol*. 2011; 66:1023–1029. [PubMed: 21861995]
41. Funama Y, Taguchi K, Utsunomiya D, Oda S, Yanaga Y, Yamashita Y. Combination of a low-tube-voltage technique with hybrid iterative reconstruction (iDose) algorithm at coronary computed tomographic angiography. *J Comput Assist Tomogr*. 2011; 35:480–485. [PubMed: 21765305]
42. Gervaise A, Osemont B, Lecocq S, Noel A, Micard E, Felblinger J. CT image quality improvement using adaptive iterative dose reduction with wide-volume acquisition on 320-detector CT. *Eur Radiol*. 2012; 22:295–301. [PubMed: 21927791]
43. Vorona GA, Ceschin RC, Clayton BL, Sutcliffe T, Tadros SS, Panigraphy A. Reducing abdominal CT radiation dose with the adaptive statistical iterative reconstruction technique in children: a feasibility study. *Pediatr Radiol*. 2011; 41:1174–1182. [PubMed: 21594551]
44. Coursey C, Frush DP, Yoshizumi T, Toncheva G, Nguyen G, Greenberg SB. Pediatric chest MDCT using tube current modulation: effect on radiation dose with breast shielding. *AJR*. 2008; 190:244. [web]W54–W61.
45. Gunn ML, Kanal KM, Kolokythas O, Anzai Y. Radiation dose to the thyroid gland and breast from multidetector computed tomography of the cervical spine: does bismuth shielding with and without a cervical collar reduce dose? *J Comput Assist Tomogr*. 2009; 33:987–990. [PubMed: 19940673]
46. American Association of Physicists in Medicine. AAPM position statement on the use of bismuth shielding for the purpose of dose reduction in CT scanning. American Association of Physicists in

Medicine website. www.aapm.org/org/policies/details.asp?id=319&type=PP. Published February 7, 2012.

47. Haliburton SS, Abbara S, Chen MY, Gentry R, Mahesh M, Raff GL. SCCT guidelines on radiation doses and dose-optimization strategies in cardiovascular CT. *J Cardiovasc Comput Tomogr*. 2011; 5:198–224. [PubMed: 21723512]
48. Jin KN, Park EA, Shin CL, Lee W, Chung JW, Park JH. Retrospective versus prospective ECG-gated dual source CT in pediatric patients with congenital heart disease: comparison of image quality and radiation dose. *Int J Cardiovasc Imaging*. 2010; 26:63–73. [PubMed: 20044793]
49. Paul JF, Rohnean A, Elfassy E, Sigal-Cinqualbre A. Radiation dose for thoracic and coronary step-and-shoot CT using a 128-slice dual-source machine in infants and small children with congenital heart disease. *Pediatr Radiol*. 2011; 41:244–249. [PubMed: 20821005]

Fig. 1.
Sample advertisement used for educational and awareness campaign conducted by Alliance for Radiation Safety in Pediatric Imaging, a 13-member organization consisting of leading medical societies, agencies, and regulatory groups that joined forces to impact patient care and change practice.

Fig. 2. Influence of tube voltage (peak kilovoltage) on absorbed dose to head and body phantoms. When tube current is fixed, dose will decrease more than linearly as tube voltage is decreased. For fixed scanner technique, absorbed dose is higher in smaller head phantom than in body phantom. CTDI_w = weighted CT dose index [26].

Fig. 3. Influence of tube current on absorbed dose to head and body phantoms. Absorbed dose is linear with tube current. CTDI_w = weighted CT dose index [26].

Fig. 4.

Influence of pitch on absorbed dose to head and body phantoms. With all other acquisition parameters fixed, absorbed dose will decrease with increasing pitch. This is result of total exposure time: scan with pitch of 2 will require 1/2 time (1/2 dose) as scan with pitch of 1. CTDI_{vol} = volume CT dose index [26].

Fig. 5.
Effect on absorbed dose to head and body phantoms depending on collimation chosen.
 $CTDI_w$ = weighted CT dose index [26].

A **B**

Fig. 6.
Illustrations of overbeaming and overranging.

A, In overbeaming, dose profiles for 4- and 16-MDCT scanners show that over same axial scan range, penumbra in 4-MDCT system will contribute to more wasted dose than from 16-MDCT system.

B, In overranging, helical acquisition requires data from approximately one half rotations before and after scan region of interest. Black section represents wasted dose for gray-shaded region of interest.

Fig. 7.

Tube current (milliamperes) superimposed on CT projection radiograph shows variation in tube current as function of time (table position along z-axis) at helical CT in 6-year-old child. Adult scanning protocol and automatic exposure control system (CareDose 4D, Siemens Healthcare) were used with reference effective tube current–time product of 165 mAs. Mean effective tube current–time product for actual scanning was 38 mAs (effective tube current–time product = tube current–time product / pitch). (Reprinted with permission from [34])

Fig. 8.

Representation of retrospective and prospective cardiac gating. Retrospective gating will subdivide tomographic data into cardiac cycles after acquisition. Most systems support modulated tube current during acquisition to lower current during systole. Prospective gating turns on x-ray tube only during one phase of cardiac cycle (usually diastole), leading to lower total dose but requiring consistent and relatively low heart rates. ECG wave-form plotted in red shows systolic gating.

TABLE 1

Adjustable Scan Parameters and Their Effect on Radiation Dose

Parameter	Effect on Radiation Dose
X-ray beam energy	Higher energy increases radiation dose (at matched tube current)
Tube current	Higher tube current increases radiation dose
Gantry rotation	Faster gantry rotation decreases radiation dose
Section thickness	Thinner collimation is linked with increased dose
Pitch	Higher pitch decreases radiation dose (at matched tube current)
Distance of x-ray tube to CT isocenter	Optimal patient placement decreases radiation dose
Scan length	Lengthening the scan range increases radiation dose